A Simple and Effective Decoupling Circuit for Three-RF-Coil ASL MRI Experiments

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Introduction

MRI for arterial spin labeling (ASL) perfusion studies involves three steps: spin labeling, spatial encoding, and data acquisition. (1) Accordingly, there are three RF events occurring sequentially: spin-labeling pulse, imaging pulse(s) and signal detection. Fig. 1 shows symbolically the timing of these events. For *in-vivo* animal experiments, it is desirable to have three separate coils to achieve best results. Clearly, success with this three-coil setup can only be achieved if the interactions between the RF coils are eliminated. In this work, we propose a decoupling scheme using RF switching property of PIN diodes.

Methods and Materials

The ideal decoupling requires that in each phase of the ASL perfusion MR experiment only one RF coil is on. This goal can be achieved by the logic shown in Fig. 1. For the labeling and transmit coils, the control logic signals come from RF blanking pulses generated automatically when RF pulses are transmitted from F1 and F2 channels which are built-in in most modern MR instruments. The logic for the receive-only coil can be

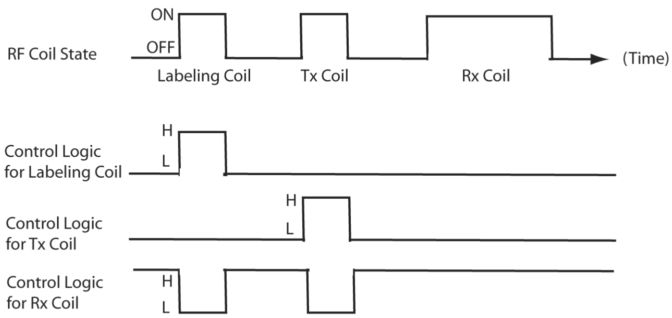

Fig.1: The timing and logic for turning on each RF coil.

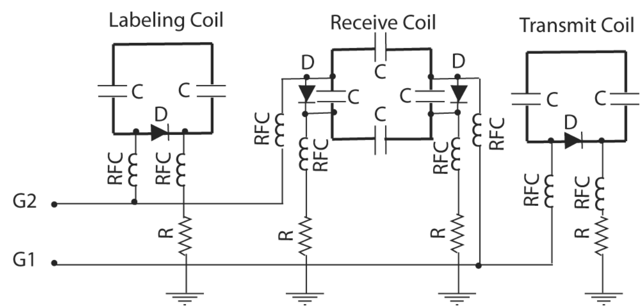

Fig.2: Schematic for three RF coil connection.

implemented by a circuit shown in Fig.2. G1 and G2 represent RF blanking pulses from channels F1 and F2, respectively. RF chokes (RFC) are used to block RF current pathways. A resistor (R) is used to limit a DC current passing through the diode. By examining Fig.2, the different combinations of high (H) and low (L) logic signals of G1 and G2 allow only one coil ON at a time. Unlike the labeling and transmit coils, the diodes (D) are not inserted in the loop of the receive coil. For this latter coil, “OFF” means detuning. In practice, such a detuning design makes the receive coil have a much higher Q factor when it is in ON state.

All MR experiments were performed on a 4.7T Bruker BioSpec imaging spectrometer operating at 200.3 MHz for ^{1}H frequency. A volume coil (Model T5346,Bruker BioSpin) that can be actively turned on or off by a DC driver was used for transmit only, the labeling coil was made of a flat, rectangular loop (3mm$\times$18mm) of copper wire, combined with a balanced matching network (2). A diode (Model T68322, Bruker BioSpin) was inserted into the loop. The receive coil was made of a flat, circular loop of copper layer etched on a printed circuit board with four ceramic chip capacitors (ATC Corp., Huntington, NY) equal distantly inserted into the loop. The loop was 18mm in dia. and 2mm in width. One of the chip capacitor was chosen as an output port and connected to a balanced matching network (2). Two diodes were connected to two different chip capacitors shown in Fig. 2. All the RF chokes and resistors were nonmagnetic. The logic control pulses were provided with an active RF decoupling box (Model T7028, Bruker BioSpin MRI Gmbh) which is a standard device attached to the MR instrument. The four logic outputs on the decoupling box are controlled by the F1 and F2 channels’ RF blanking pulses. The box outputs +5V as logic high and –32V as logic low.

All RF coils were first evaluated separately on the bench. The logic signal was provided with a DC power supply. The frequency characteristic was measured with a HP network/spectrum analyzer (HP 4195A, HP Company, Rockville, IL). The “ON” or “OFF” of the diodes was achieved by +5V or –32V corresponding to high or low logic level, respectively. For the “ON” labeling coil, the Q value was somewhat lower compared to that of the coil without the diode. However, for the “ON” receive coil, the Q values were virtually unchanged with or without the diodes.

Subsequently, the three-coil setup was evaluated with MR imaging on a water phantom. An ASL perfusion sequence (1) was used with labeling pulses delivered through F2 channel and imaging pulses delivered through F1 channel. 2mm-thick slice in a transverse plane was imaged using spin-echo sequence with the following parameters: FOV=25mm, matrix=128$\times$128, TR=1s, TE 10ms, NA=1. Two images, with and without the labeling pulse, were collected.The three-coil approach was evaluated next *in vivo*, on a rat. The RF power driving the labeling coil was calibrated using a previously reported method (1). The perfusion imaging was performed with a spin-echo sequence and the following parameters. FOV=50mm, matrix=128$\times$128, slice thickness=2mm, TR=850ms, TE=10ms, length for a labeling pulse=750ms.

Results and Discussion

No changes in the resonant frequency value and shape were observed for each of the three coils connected according to the scheme shown in Fig.2, as compared to their resonant characteristics measured individually. This indicates that only one coil is turned on as long as the control logic given in Fig.1 is applied. Since the water was stationary, the two images, acquired on the water phantom, are virtually identical, as confirmed by the clean difference images between these two images. The uniform image intensity also demonstrates that the interactions between any of the three RF coils are completely eliminated. The perfusion images acquired with the three-coil setup, *in vivo*, on the rat brain also show good contrast and no artifacts.

Conclusion

We have proposed and implement a simple circuit that connects labeling, transmit-only and receive-only coils used for ASL perfusion MRI experiments. Both the bench measurements and MRI results show that the interactions between the RF coils are effectively eliminated with the proposed circuit. (This work was supported by grant MH58912 and NS44617 from NIH.)

References

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